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Hoffman et al.

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(54) CT DETECTOR HAVING A SEGMENTED OPTICAL COUPLER AND METHOD OF MANUFACTURING SAME

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Related U.S. Application Data

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(51) **Int. Cl.**
G01T 1/24 (2006.01)

(52) **U.S. Cl.** **250/368**; 378/4; 378/98.8

(58) **Field of Classification Search** 250/362, 250/368, 370.09, 370.11
See application file for complete search history.

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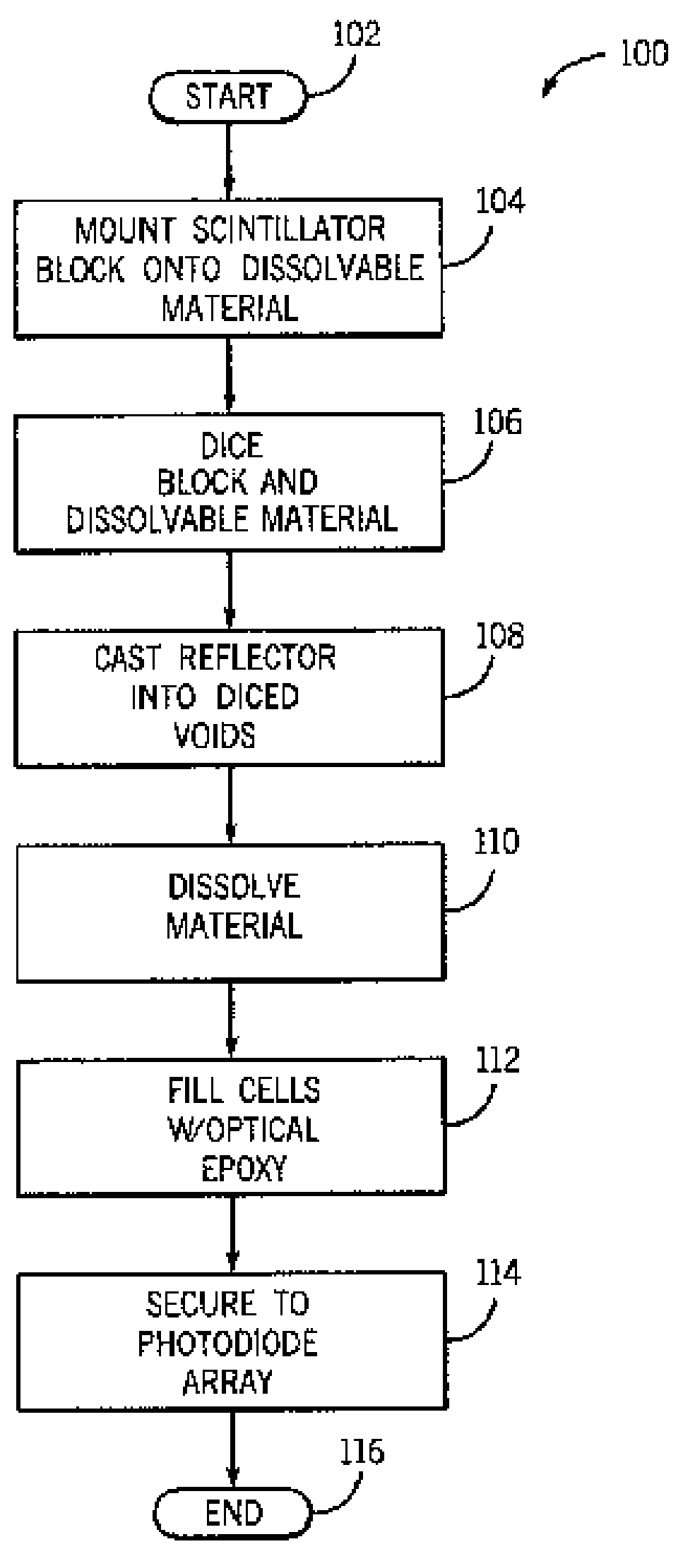
(74) *Attorney, Agent, or Firm*—Zielkowski Patent Solutions Group, SC; Michael A. Della Penna; Carl B. Horton

(57) ABSTRACT

The present invention is a directed to a CT detector for a CT imaging system that incorporates a segmented optical coupler between a photodiode array and a scintillator array. The segmented optical coupler also operates as a light collimator which improves the light collection efficiency of the photodiode array. The segmented optical coupler is defined by a series of reflector elements that collectively form a plurality of open cells. The open cells form light transmission cavities and facilitate the collimation of light from a scintillator to a photodiode. The cavities may be filled with optical epoxy for sealing to the photodiode array.

16 Claims, 9 Drawing Sheets



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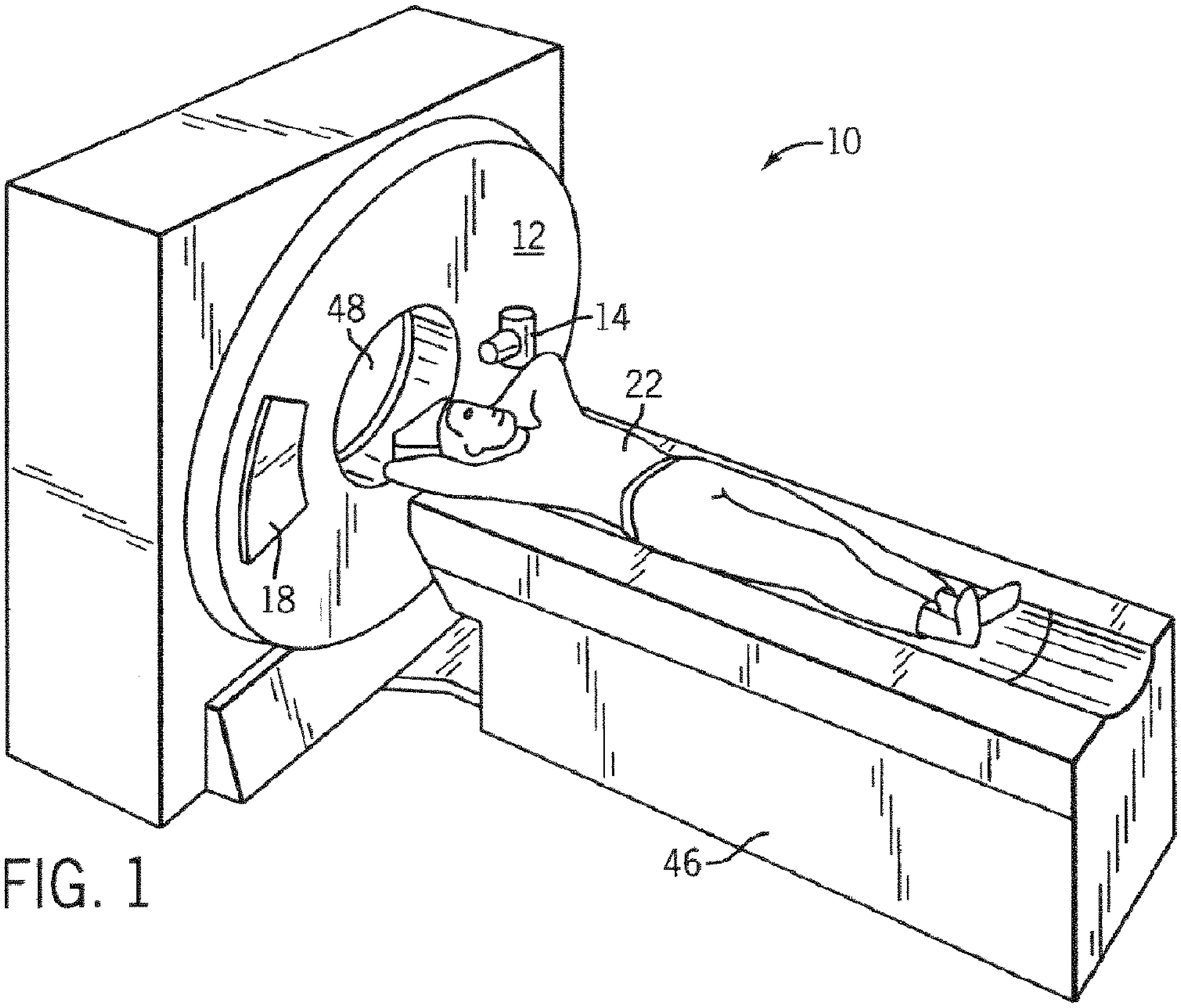
10
12
48
14
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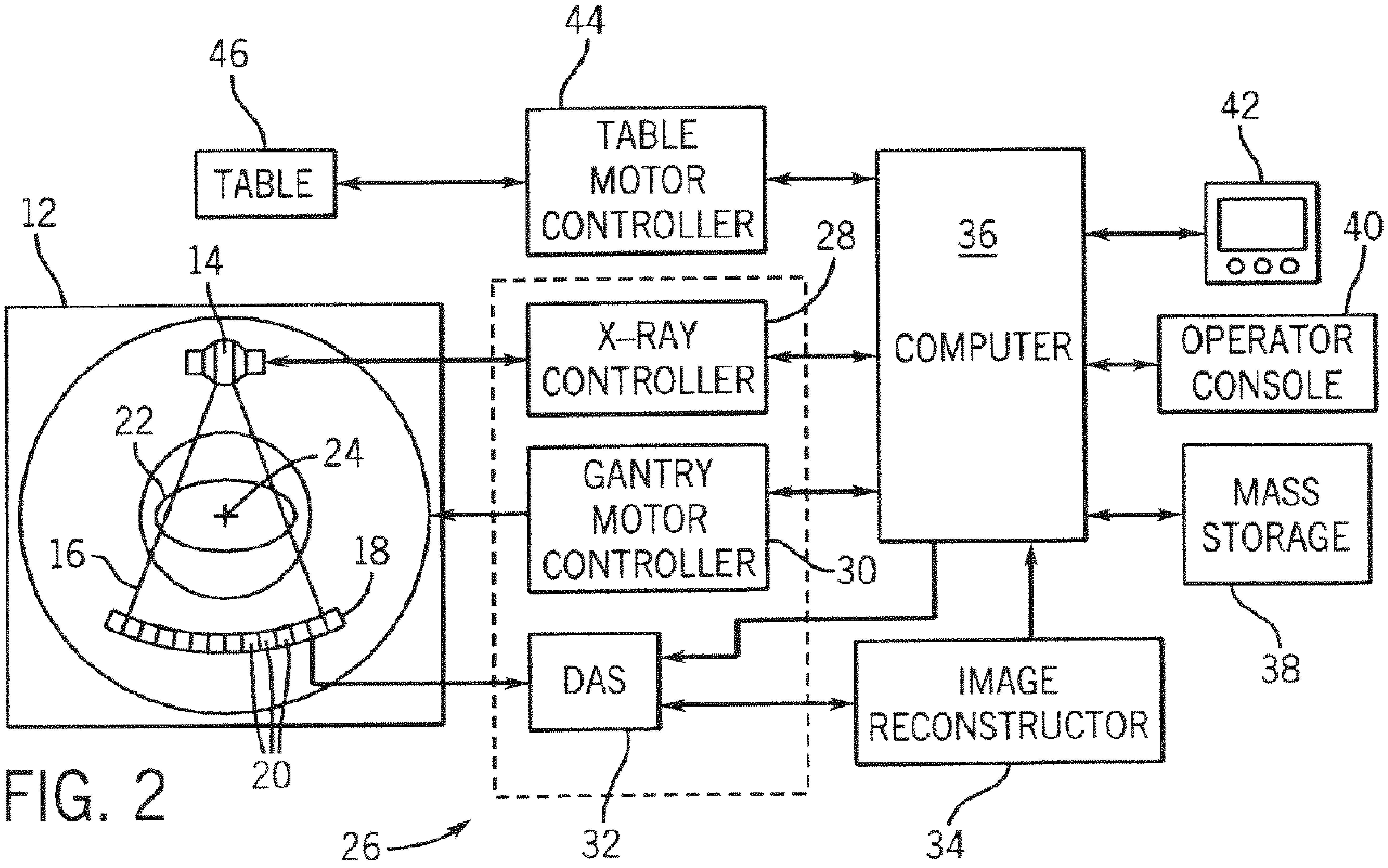
46

FIG. 1


46
TABLE
44
TABLE MOTOR CONTROLLER
36
COMPUTER
42
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OPERATOR CONSOLE
12
14
28
X-RAY CONTROLLER
22
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GANTRY MOTOR CONTROLLER
30
16
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MASS STORAGE
38
DAS
IMAGE RECONSTRUCTOR
20
26
32
34

FIG. 2

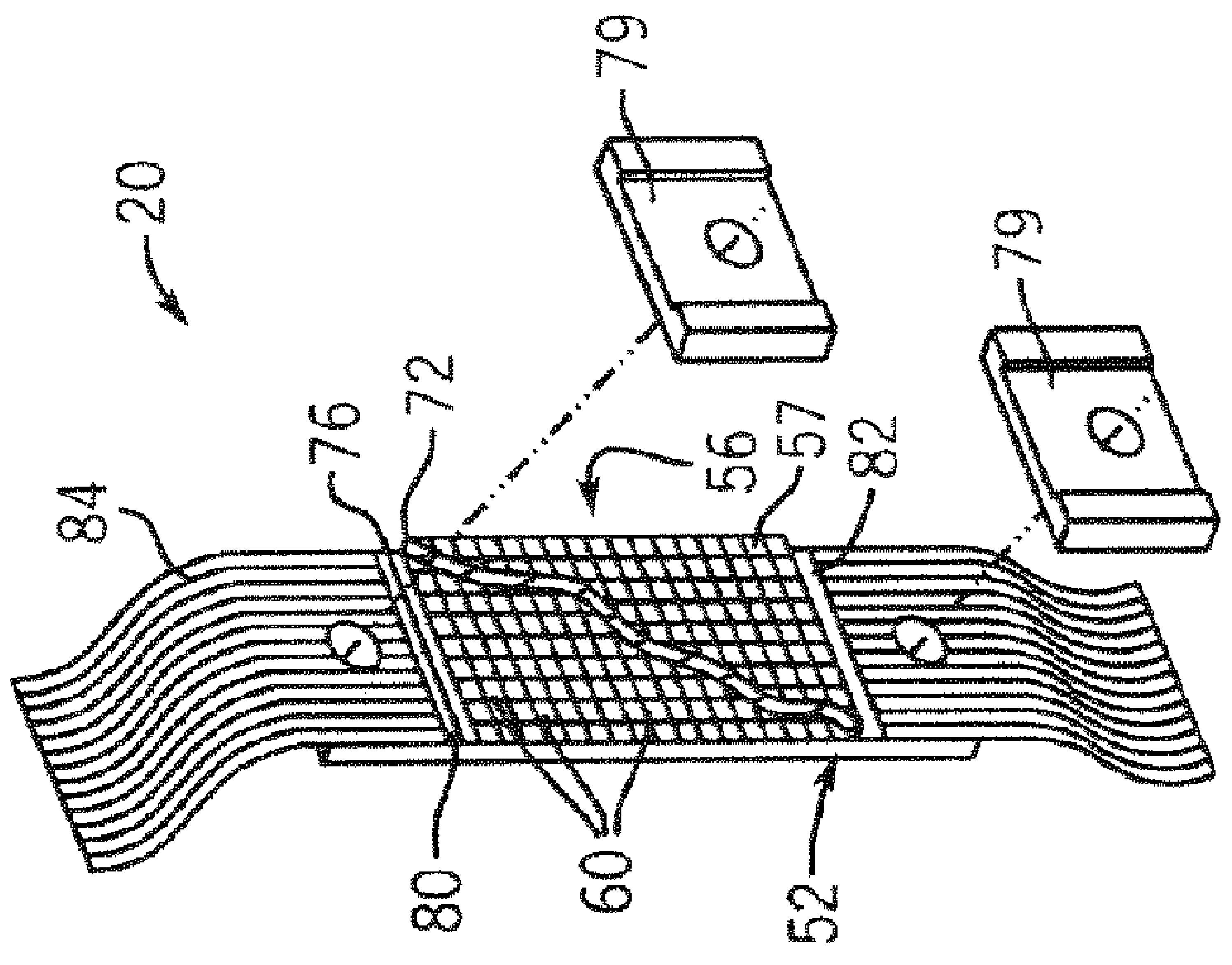


FIG. 4

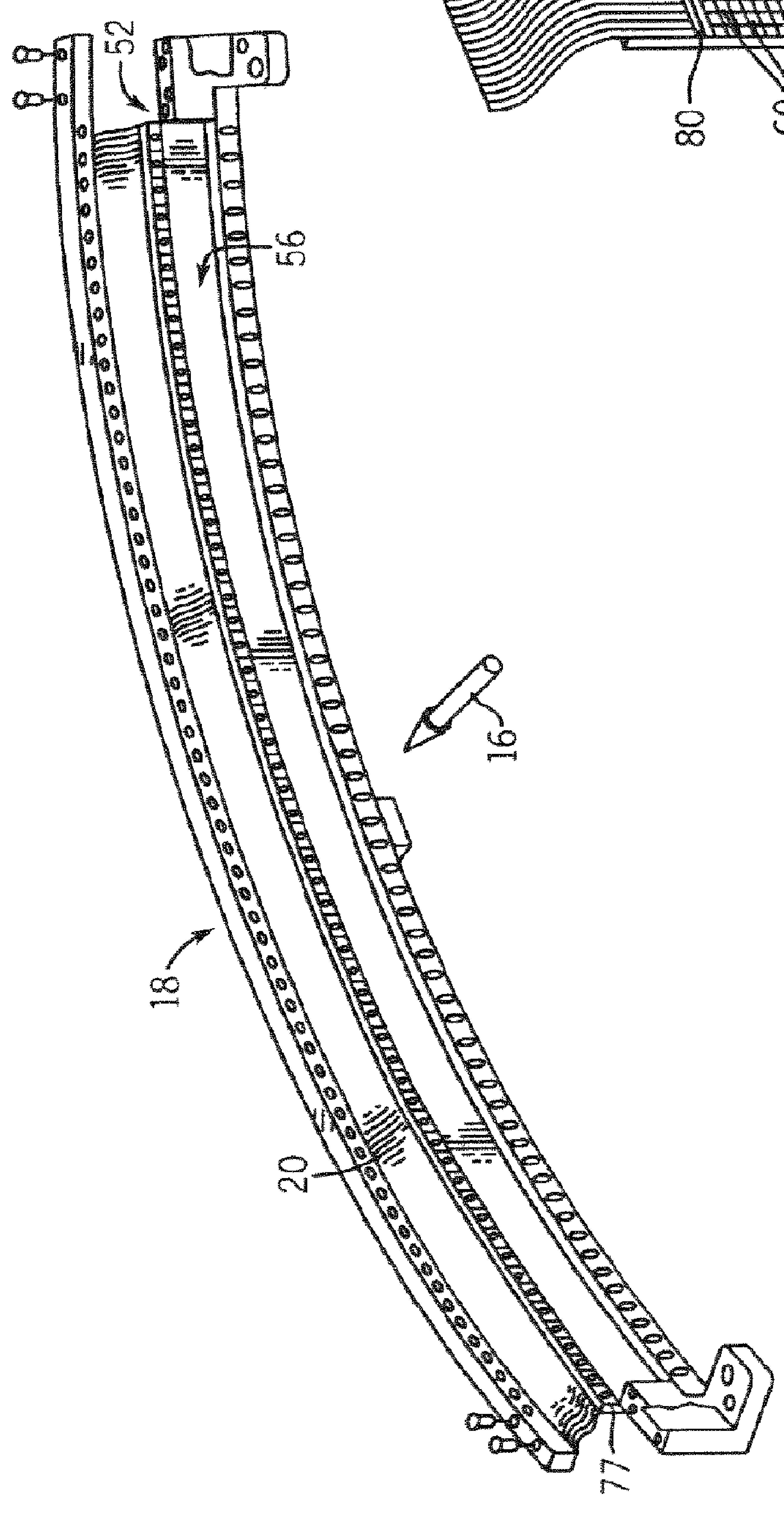
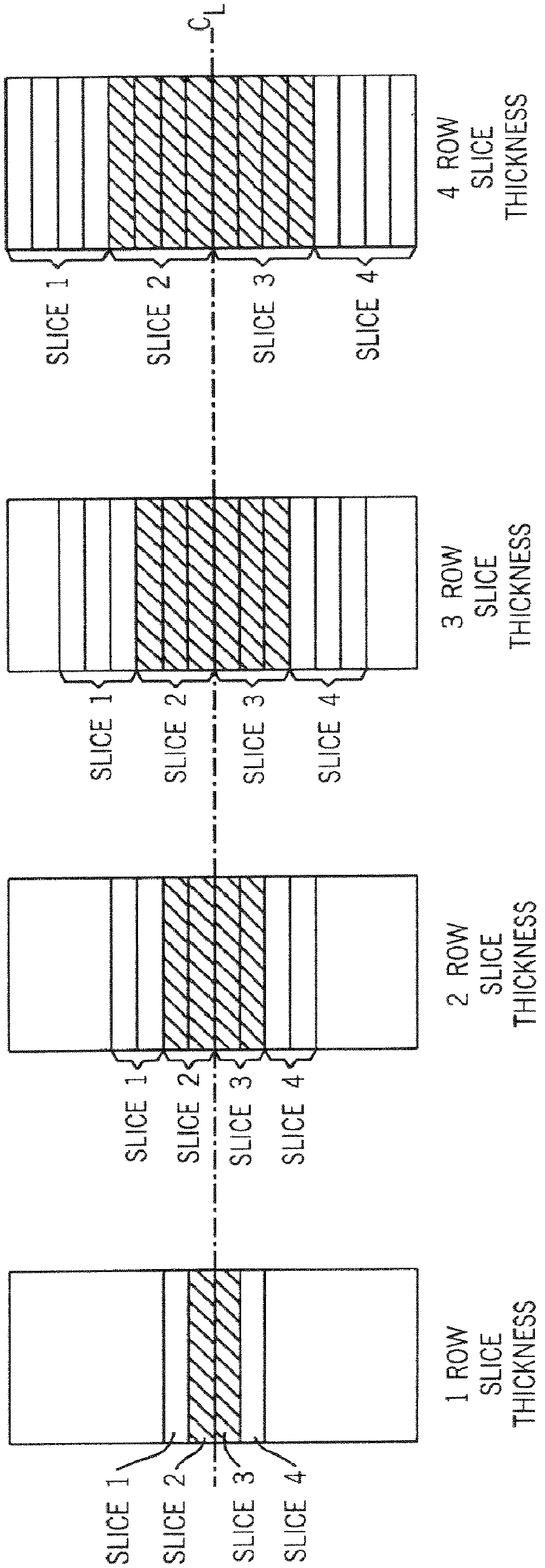


FIG. 3


SLICE 1
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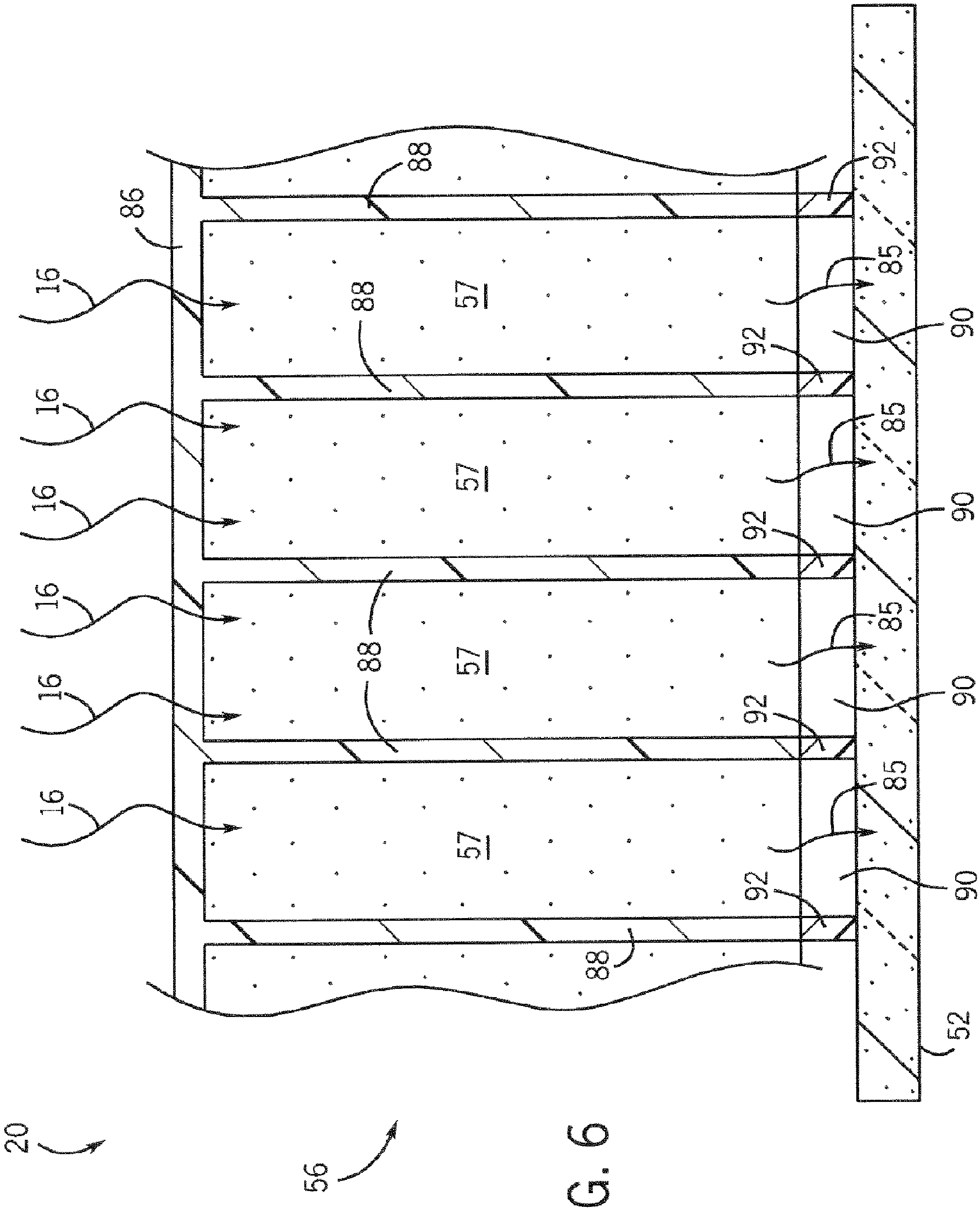
SLICE 3
SLICE 4
1 ROW SLICE THICKNESS
SLICE 1
SLICE 2
SLICE 3
SLICE 4
2 ROW SLICE THICKNESS
SLICE 1
SLICE 2
SLICE 3
SLICE 4
3 ROW SLICE THICKNESS
SLICE 1
SLICE 2
SLICE 3
SLICE 4
4 ROW SLICE THICKNESS
C_L

FIG. 5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56

FIG. 6

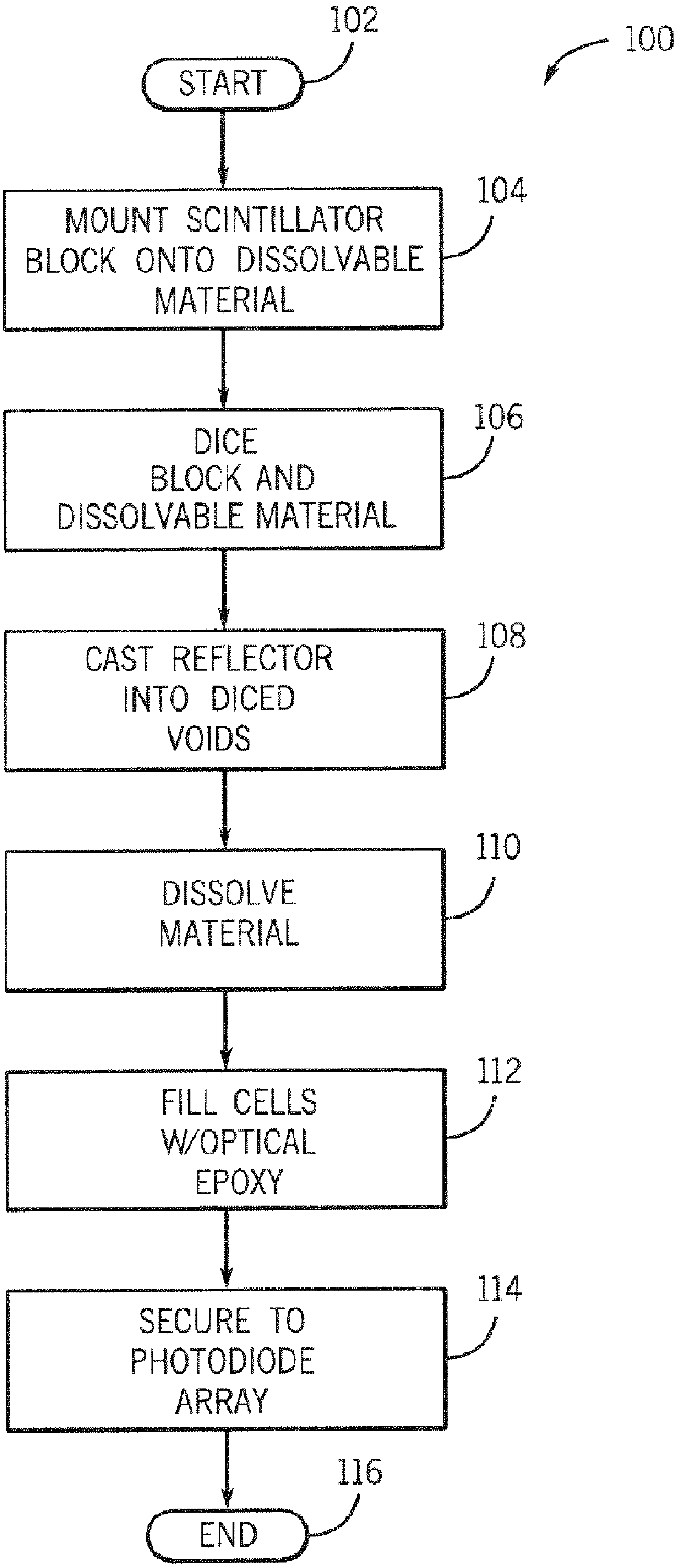
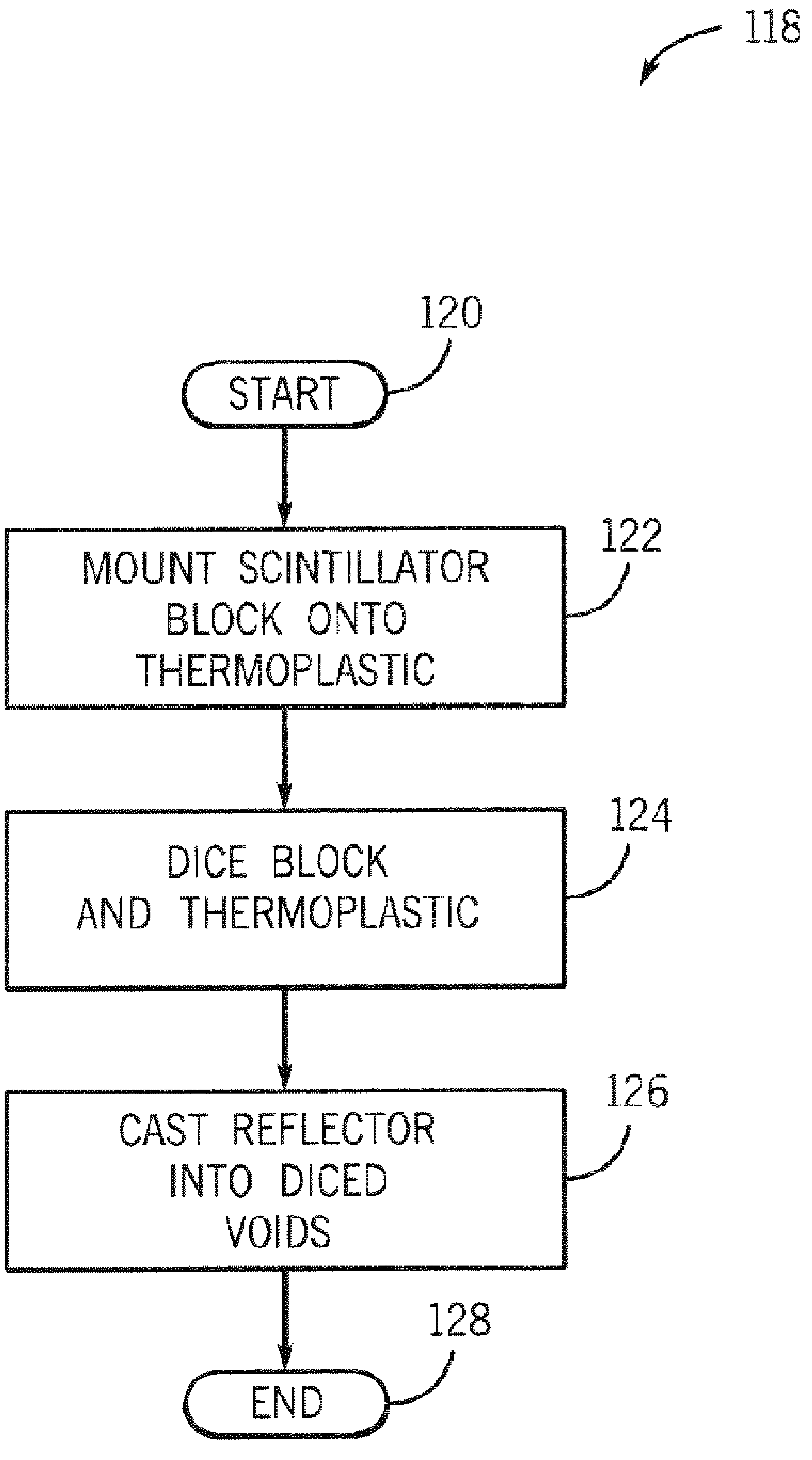

102
100
START
104
MOUNT SCINTILLATOR BLOCK ONTO DISSOLVABLE MATERIAL
106
DICE BLOCK AND DISSOLVABLE MATERIAL
108
CAST REFLECTOR INTO DICED VOIDS
110
DISSOLVE MATERIAL
112
FILL CELLS W/OPTICAL EPOXY
114
SECURE TO PHOTODIODE ARRAY
116
END

FIG. 7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120
START
MOUNT SCINTILLATOR BLOCK ONTO THERMOPLASTIC
122
DICE BLOCK AND THERMOPLASTIC
124
CAST REFLECTOR INTO DICED VOIDS
126
128
END

FIG. 8

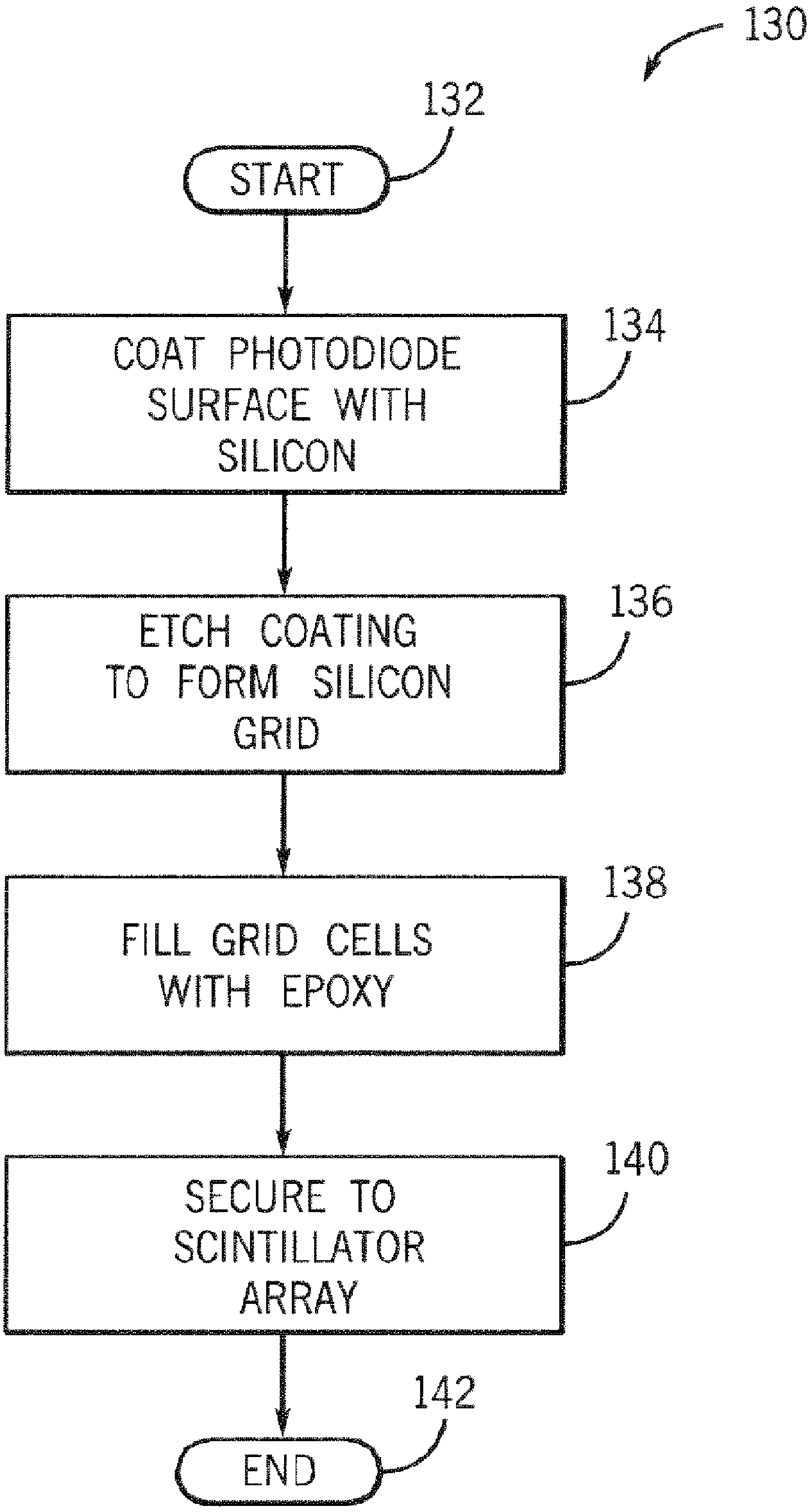
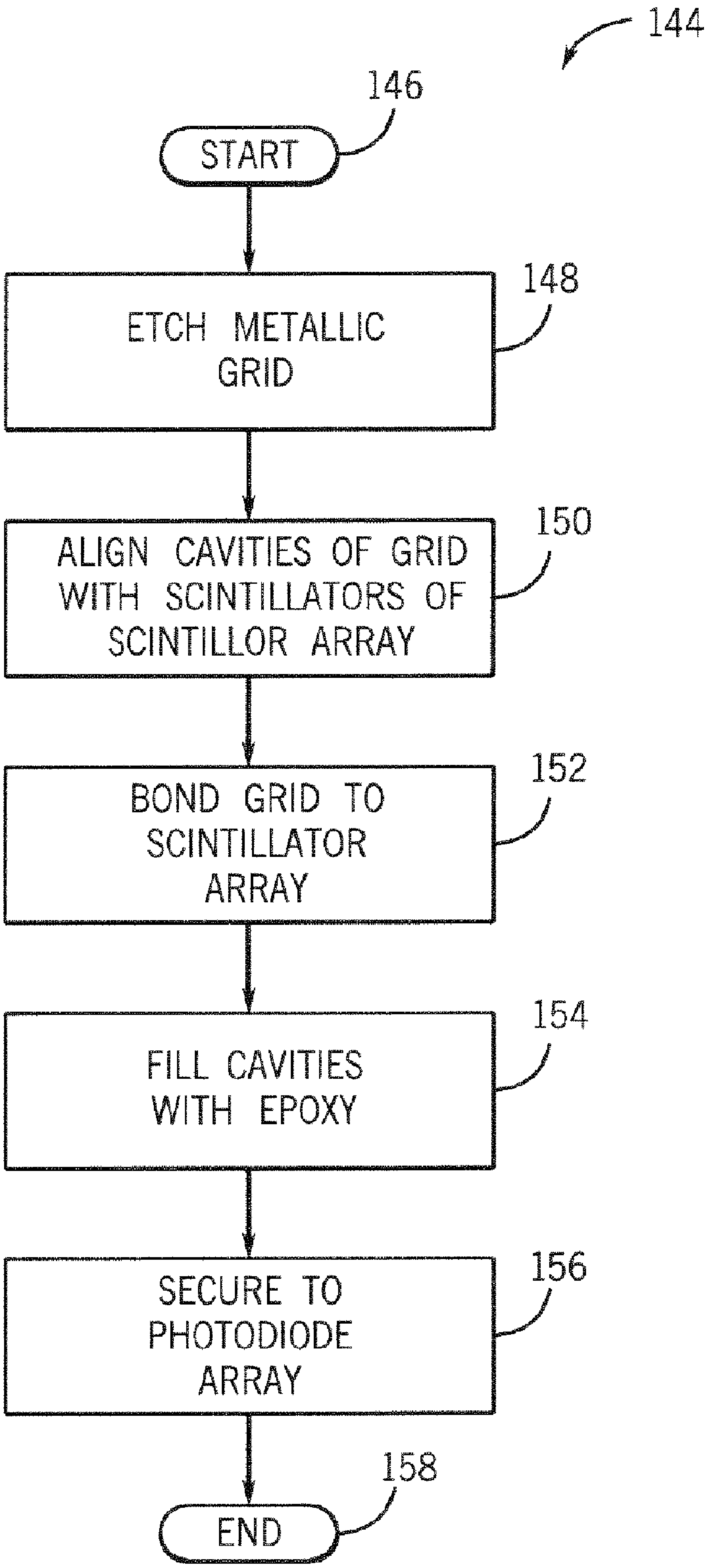

130
132
START
134
COAT PHOTODIODE SURFACE WITH SILICON
136
ETCH COATING TO FORM SILICON GRID
138
FILL GRID CELLS WITH EPOXY
140
SECURE TO SCINTILLATOR ARRAY
142
END

FIG. 9


144
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START
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ETCH METALLIC GRID
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ALIGN CAVITIES OF GRID WITH SCINTILLATORS OF SCINTILLOR ARRAY
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BOND GRID TO SCINTILLATOR ARRAY
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FILL CAVITIES WITH EPOXY
156
SECURE TO PHOTODIODE ARRAY
158
END

FIG. 10

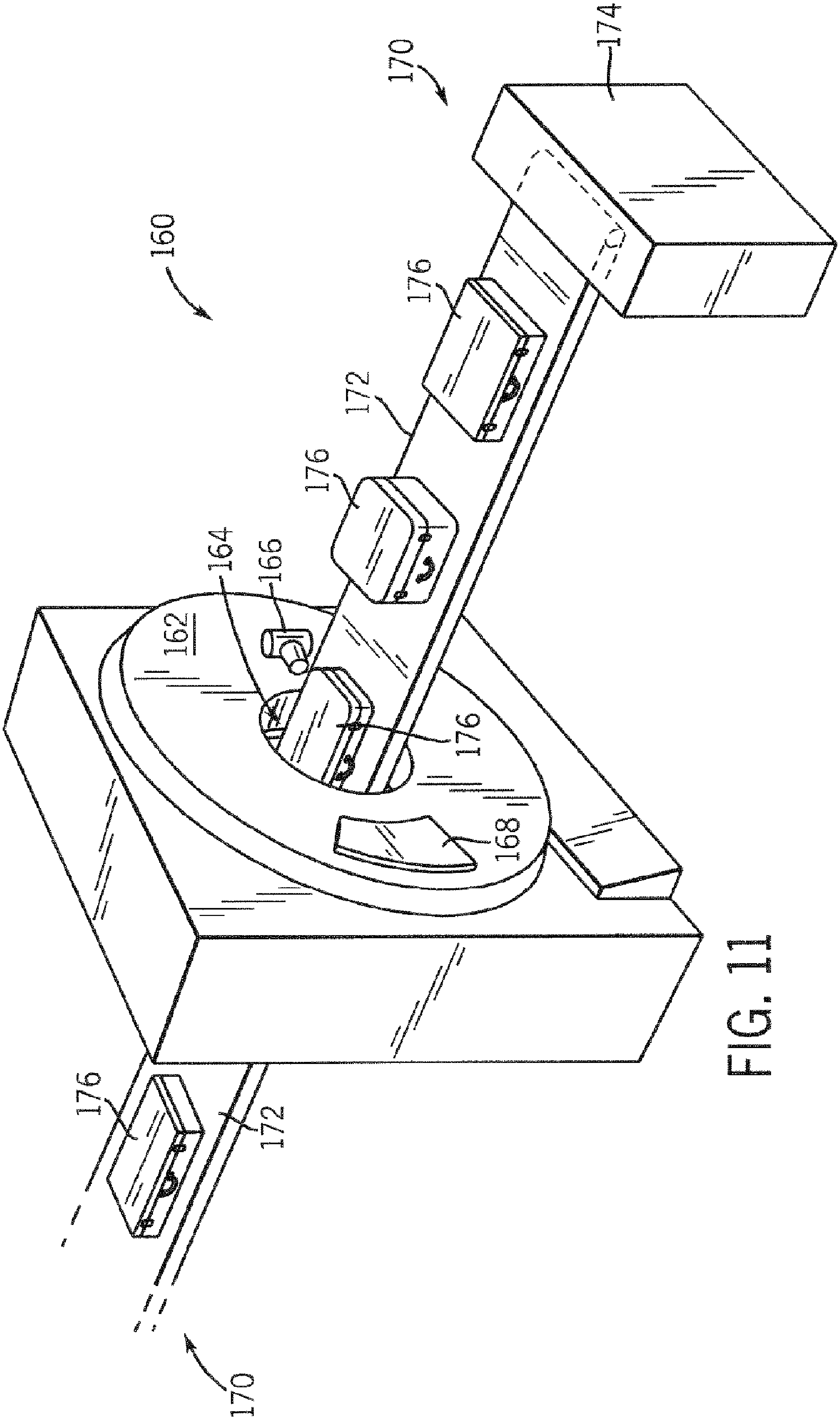

FIG. 11

CT DETECTOR HAVING A SEGMENTED OPTICAL COUPLER AND METHOD OF MANUFACTURING SAME

CROSS REFERENCE TO RELATED APPLICATIONS

The present application is divisional of U.S. Ser. No. 10/908,209 filed May 2, 2005, which is a continuation of and claims priority of U.S. Ser. No. 10/249,052 filed Mar. 12, 2003 now U.S. Pat. No. 6,933,504.

BACKGROUND OF THE INVENTION

The present invention relates generally to diagnostic imaging and, more particularly, to a CT detector having a segmented or non-contiguous optical coupler and method of manufacturing same. Additionally, the segmented optical coupler operates as a light collimator integrally formed between the scintillators and photodiodes of the detector.

Typically, in computed tomography (CT) imaging systems, an x-ray source emits a fan-shaped beam toward a subject or object, such as a patient or a piece of luggage. Hereinafter, the terms "subject" and "object" shall include anything capable of being imaged. The beam, after being attenuated by the subject, impinges upon an array of radiation detectors. The intensity of the attenuated beam radiation received at the detector array is typically dependent upon the attenuation of the x-ray beam by the subject. Each detector element of the detector array produces a separate electrical signal indicative of the attenuated beam received by each detector element. The electrical signals are transmitted to a data processing system for analysis which ultimately produces an image.

Generally, the x-ray source and the detector array are rotated about the gantry within an imaging plane and around the subject. X-ray sources typically include x-ray tubes, which emit the x-ray beam at a focal point. X-ray detectors typically include a collimator for collimating x-ray beams received at the detector, a scintillator for converting x-rays to light energy adjacent the collimator, and photodiodes for receiving the light energy from the adjacent scintillator and producing electrical signals therefrom.

Typically, each scintillator of a scintillator array converts x-rays to light energy. Each scintillator discharges light energy to a photodiode adjacent thereto. Each photodiode detects the light energy and generates a corresponding electrical signal. The outputs of the photodiodes are then transmitted to the data processing system for image reconstruction.

"Cross talk" between detector cells of a CT detector is common. "Cross talk" is generally defined as the communication of data between adjacent cells of a CT detector. Generally, cross talk is sought to be reduced as cross talk leads to artifact presence in the final reconstructed CT image and contributes to poor spatial resolution. Typically, four different types of cross talk may result within a single CT detector. X-ray cross talk may occur due to x-ray scattering between scintillator cells. Optical cross talk may occur through the transmission of light through the reflectors that surround the scintillators. Known CT detectors utilize a contiguous optical coupling layer(s), typically epoxy, to secure the scintillator array to the photodiode array. Cross talk, however, can occur as light from one cell is passed to another through the contiguous layer. Electrical cross talk can occur from unwanted communication between photodiodes. Of the above types of cross talk, cross talk though the contiguous optical coupler layer(s) is generally considered a major source of cross talk in the CT detector.

Therefore, it would be desirable to design a CT detector having improved optical coupling between the scintillator array and photodiode array to reduce cross talk in the CT detector and improve spatial resolution of the final reconstructed image.

BRIEF DESCRIPTION OF THE INVENTION

The present invention is a directed to a CT detector for a CT imaging system that overcomes the aforementioned drawbacks. The CT detector incorporates a gridded light collimator between a photodiode array and a scintillator array. The light collimator improves the light collection efficiency of the photodiode array and may be formed of reflector material so as to reduce cross talk within the detector. Each gridded collimator is defined by a series of reflector elements that collectively form a plurality of open cells. The open cells form light transmission cavities and facilitate the collimation of light from a scintillator to a photodiode. The cavities may be filled with optical epoxy for sealing to the photodiode array or scintillator array thereby avoiding the drawbacks associated with contiguous optical coupler layers.

Therefore, in accordance with the present invention, a CT detector includes a plurality of scintillators arranged in an array to receive x-rays and output light in response to the received x-rays. A plurality of light detection elements are arranged in an array dimensionally similar to the scintillator array and are configured to detect light from the scintillators. A non-contiguous optical coupler is then used to secure the plurality of scintillators to the plurality of light detection elements.

According to another aspect of the present invention, a CT system includes a rotatable gantry having a bore centrally disposed therein and a table movable fore and aft through the bore and configured to position a subject for CT data acquisition. A high frequency electromagnetic energy projection source is positioned within the rotatable gantry and configured to project high frequency electromagnetic energy toward the subject. The CT system further includes a detector array disposed within the rotatable gantry and configured to detect high frequency electromagnetic energy projected by the projection source and impinged by the subject. The detector array includes a plurality of scintillators arranged in a scintillator array as well as a plurality of photodiodes arranged in a photodiode array. A light collimator having a plurality of light transmission cavities is disposed between the scintillator array and the photodiode array.

In accordance with a further aspect of the present invention, a method of CT detector manufacturing includes the steps of forming a scintillator array having a plurality of scintillators and forming a photodiode array having a plurality of photodiodes. An open-celled collimator is then deposited between the arrays. The resulting assembly is then secured to one another.

Various other features, objects and advantages of the present invention will be made apparent from the following detailed description and the drawings.

BRIEF DESCRIPTION OF DRAWINGS

The drawings illustrate one preferred embodiment presently contemplated for carrying out the invention.

In the drawings:

FIG. 1 is a pictorial view of a CT imaging system.

FIG. **2** is a block schematic diagram of the system illustrated in FIG. **1**.

FIG. **3** is a perspective view of one embodiment of a CT system detector array.

FIG. **4** is a perspective view of one embodiment of a detector.

FIG. **5** is illustrative of various configurations of the detector in FIG. **4** in a four-slice mode.

FIG. **6** is a schematic of a cross-section of a CT detector in accordance with the present invention.

FIGS. **7**–**10** set forth steps of various techniques of manufacturing a CT detector in accordance with the present invention.

FIG. **11** is a pictorial view of a CT system for use with a non-invasive package inspection system.

DETAILED DESCRIPTION

The operating environment of the present invention is described with respect to a four-slice computed tomography (CT) system. However, it will be appreciated by those skilled in the art that the present invention is equally applicable for use with single-slice or other multi-slice configurations. Moreover, the present invention will be described with respect to the detection and conversion of x-rays. However, one skilled in the art will further appreciate that the present invention is equally applicable for the detection and conversion of other high frequency electromagnetic energy. The present invention will be described with respect to a "third generation" CT scanner, but is equally applicable with other CT systems.

Referring to FIGS. **1** and **2**, a computed tomography (CT) imaging system **10** is shown as including a gantry **12** representative of a "third generation" CT scanner. Gantry **12** has an x-ray source **14** that projects a beam of x-rays **16** toward a detector array **18** on the opposite side of the gantry **12**. Detector array **18** is formed by a plurality of detectors **20** which together sense the projected x-rays that pass through a medical patient **22**. Each detector **20** produces an electrical signal that represents the intensity of an impinging x-ray beam and hence the attenuated beam as it passes through the patient **22**. During a scan to acquire x-ray projection data, gantry **12** and the components mounted thereon rotate about a center of rotation **24**.

Rotation of gantry **12** and the operation of x-ray source **14** are governed by a control mechanism **26** of CT system **10**. Control mechanism **26** includes an x-ray controller **28** that provides power and timing signals to an x-ray source **14** and a gantry motor controller **30** that controls the rotational speed and position of gantry **12**. A data acquisition system (DAS) **32** in control mechanism **26** samples analog data from detectors **20** and converts the data to digital signals for subsequent processing. An image reconstructor **34** receives sampled and digitized x-ray data from DAS **32** and performs high speed reconstruction. The reconstructed image is applied as an input to a computer **36** which stores the image in a mass storage device **38**.

Computer **36** also receives commands and scanning parameters from an operator via console **40** that has a keyboard. An associated cathode ray tube display **42** allows the operator to observe the reconstructed image and other data from computer **36**. The operator supplied commands and parameters are used by computer **36** to provide control signals and information to DAS **32**, x-ray controller **28** and gantry motor controller **30**. In addition, computer **36** operates a table motor controller **44** which controls a motorized table **46** to position patient **22** and gantry **12**. Particularly, table **46** moves portions of patient **22** through a gantry opening **48**.

As shown in FIGS. **3** and **4**, detector array **18** includes a plurality of scintillators **57** forming a scintillator array **56**. A collimator (not shown) is positioned above scintillator array **56** to collimate x-ray beams **16** before such beams impinge upon scintillator array **56**.

In one embodiment, shown in FIG. **3**, detector array **18** includes **57** detectors **20**, each detector **20** having an array size of 16×16. As a result, array **18** has 16 rows and 912 columns (16×57 detectors) which allows 16 simultaneous slices of data to be collected with each rotation of gantry **12**.

Switch arrays **80** and **82**, FIG. **4**, are multi-dimensional semiconductor arrays coupled between scintillator array **56** and DAS **32**. Switch arrays **80** and **82** include a plurality of field effect transistors (EFT) (not shown) arranged as multi-dimensional array. The EFT array includes a number of electrical leads connected to each of the respective photodiodes **60** and a number of output leads electrically connected to DAS **32** via a flexible electrical interface **84**. Particularly, about one-half of photodiode outputs are electrically connected to switch **80** with the other one-half of photodiode outputs electrically connected to switch **82**. Additionally, a reflector layer (not shown) may be interposed between each scintillator **57** to reduce light scattering from adjacent scintillators. Each detector **20** is secured to a detector frame **77**, FIG. **3**, by mounting brackets **79**.

Switch arrays **80** and **82** further include a decoder (not shown) that enables, disables, or combines photodiode outputs in accordance with a desired number of slices and slice resolutions for each slice. Decoder, in one embodiment, is a decoder chip or a FET controller as known in the art. Decoder includes a plurality of output and control lines coupled to switch arrays **80** and **82** and DAS **32**. In one embodiment defined as a 16 slice mode, decoder enables switch arrays **80** and **82** so that all rows of the photodiode array **52** are activated, resulting in 16 simultaneous slices of data for processing by DAS **32**. Of course, many other slice combinations are possible. For example, decoder may also select from other slice modes, including one, two, and four-slice modes.

As shown in FIG. **5**, by transmitting the appropriate decoder instructions, switch arrays **80** and **82** can be configured in the four-slice mode so that the data is collected from four slices of one or more rows of photodiode array **52**. Depending upon the specific configuration of switch arrays **80** and **82**, various combinations of photodiodes **60** can be enabled, disabled, or combined so that the slice thickness may consist of one, two, three, or four rows of scintillator array elements **57**. Additional examples include, a single slice mode including one slice with slices ranging from 1.25 mm thick to 20 mm thick, and a two slice mode including two slices with slices ranging from 1.25 mm thick to 10 mm thick. Additional modes beyond those described are contemplated.

Referring now to FIG. **6**, a schematic of a cross-section of a CT detector **20** is shown. As discussed above, detector **20** includes a scintillator array **56** defined by a plurality of scintillators **57**. Each of the scintillators is designed to generate a light output **85** in response the reception of x-rays **16**. A reflector layer **86** coats the x-ray reception surface of the scintillators to improve light collection efficiency of the photodiodes. The reflector layer **86** is composed of a material that allows x-rays projected from a projection source to pass through and reflects light generated by the scintillators back toward the photodiodes. The reflector layer is integrated with a series of reflector elements **88** that extend between adjacent scintillators **57** as a reflector wall. The reflector elements **88** are designed to prevent light scattering and/or reduce x-ray scattering between scintillators.

CT detector **20** is constructed such that a light cavity **90** extends between each photodiode and scintillator. The light cavity may be constructed in accordance with a number of fabrication techniques as will be described with respect to FIGS. **7**–**10** and is defined by cavity elements or plates **92**. Plates **92** are preferably formed of a reflector material similar to that used to form reflector elements **88**. Additionally, plates **92** have a width similar to the width of the reflector elements **88**. Preferably, plates **92** are formed during the formation of reflector elements **88**, as will be described with respect to FIG. **7**. As such, plates **92** extend from reflector elements to the light detection surface of the photodiode array.

Plates **92** are constructed to form light transmission cavities **90** and, as such, operate as an inner-cell light collimator. Plates **92** are designed to eliminate light cross talk between scintillators thereby collimating light toward the light detection surfaces of the photodiode array. Further, plates **92** may be coated with an optical coupling film or resin so as to secure the plates to the photodiode array. Alternately, the plates may be bonded to the surface of the photodiode array. In a further embodiment, each of the light transmission cavities **90** is filled with an optical epoxy similar to the epoxy used in a contiguous epoxy layer. The optical epoxy operates as adhesive to connect the photodiode array to the scintillator array. With the presence of reflector plates **92**, the drawbacks associated with contiguous optical layer cross talk are avoided. While epoxy may be used to secure the arrays to one another, other composites and materials such as thermoplastics may be used and are within the scope of the invention.

Referring now to FIG. **7**, steps for a technique of manufacturing a CT detector similar to that described with respect to FIG. **6** are shown. The steps illustrated may be carried out by a labor intensive process, a fully automated, computer driven process, or a combination thereof. Technique **100** begins at **102** with the assimilation of products, personnel and the like for CT detector fabrication. That achieved during this step may vary but, at a minimum, should include the preparation of a scintillator block. The scintillator block is then mounted onto a dissolvable material **104**. The scintillator block and dissolvable material are then diced or cut at **106**. Once cut, either along one or two dimensions, a plurality of scintillator cells uniformly spaced from one another results. Reflector material is then cast at **108** in the voids created between the scintillator cells as a result of the dicing process. The reflector material should be cast such that the interface between scintillators is completely filled as is the interface between adjacent portions of the dissolvable material. The cast reflector material is then allowed to cure and undergoes any additional processing to insure proper reflectivity and the like. Once the cast reflector material has cured, the dissolvable material is dissolved at **110**. The process for dissolving the material depends on the type of dissolvable material used. For example, the dissolvable material may be placed in a wash and chemically dissolved or heated at a specified temperature to, in essence, "melt" away the dissolvable material. After the dissolving process is complete, a scintillator array with an integrated cast reflector results. Of particular note is that each reflector element between the scintillators extends beyond the scintillator, i.e. has a greater length than the scintillators. The portion of the reflector that extends beyond the scintillator operates as a reflector plate as described above. The open cells that result between reflector plates define a light transmission cavity and are filled with optical epoxy at **112**. The optical epoxy permits the transmission of light between scintillator and photodiode while simultaneously creating an adhesion interface for coupling the scintillator to the photodiode. As such, the photodiode array and scintillator array are coupled to one another at **114**. This portion of the CT detector fabrication process is then complete and the remainder of the CT detector fabricating takes place downstream at **116**.

The CT detector described with respect to FIG. **6** and fabricated according to the technique of FIG. **7** illustrates only one example of the present invention. A similar CT detector incorporating the advantages of that described with respect to FIG. **6** and fabricated in accordance with techniques different from that illustrated in FIG. **7** are contemplated and within the scope of this invention. For purpose of illustration and not limitation, additional manufacturing techniques and the resulting structures will be described with reference to FIGS. **8**–**10**.

Referring now to FIG. **8**, another CT manufacturing process **118** begins at **120** with a block of scintillator material being prepared. The block is then placed onto a block of thermoplastic material at **122**. The scintillator block and the thermoplastic are then diced or cut **124** in accordance with known dicing processes. Preferably, only a portion of the thermoplastic is diced thereby leaving a thin, uncut portion that can be used to seal against the photodiode array. Cast reflector is then deposited in the voids **126** between scintillator cells that result from the dicing process. In contrast to the CT detector constructed in accordance with FIG. **7**, an optical epoxy between the reflector plates formed by the cast reflector is not used. Because the thermoplastic material is not completely diced through, a thin thermoplastic layer results that, as discussed above, is used to secure the scintillator array to the photodiode array as opposed to an optical epoxy. Process **118** then concludes at **128** with the CT detector undergoing additional processing and fabrication in accordance with known techniques.

The processes described above involve alternations to the scintillator array. In contrast, the process of FIG. **9** creates the reflector plates by etching the photodiode array. Specifically, process **130** begins at **132** with the formation of a photodiode array. At **134**, the photodiode array is coated with a film of semiconductor or other suitable materials. Preferably, a thin layer of Silicon is applied or thermally grown and allowed to cure to the photodiode light reception surface. Semiconductor materials that will not adversely affect the light collection abilities of the photodiode array should be used. The surface of the photodiode array is then masked and plasma etched at **136** using standard semiconductor fabrication techniques to form a grid. Various semiconductor fabrication processes are contemplated including chemical etching, mechanical etching, ion beam milling, and the like. The result of the etching process should result in a series of open cells defined by the semiconductor material. The open cells should be vertically aligned with the light detection surfaces of the photodiode array. The open cells are then filled with optical epoxy **138** to secure the photodiode array to the scintillator array at **140**. The resulting assembly then undergoes standard post-processing techniques whereupon the process ends at **142**.

The process illustrated in FIG. **10** utilizes an intermediary element that is not integrated with the scintillator array or photodiode array. Manufacturing process **144** begins at **146** with the formation of a scintillator array and a photodiode array in accordance with known fabrication techniques. A grid is then etched at **148** from a sheet of thin metallic or other material. The grid defines a number of cells dimensionally equivalent to the scintillators and photodiodes. Additionally, the grid preferably has a height equal to the desired height of the light transmission cavities heretofore described. Accordingly, the open cells formed in the grid are aligned with the either the scintillators of the scintillator array or with the photodiodes of the photodiode array at **150**. The grid is then bonded at **152** to the selected array. The open cells or cavities defined by the grid may then be filled with optical epoxy at **154**. The optical epoxy is then used to secure the selected array to the other array at **156**. Alternately, the open cells may be left empty and the grid bonded to the other array. The process is then complete at **158**.

Each of the above-described manufacturing processes results in a CT detector having a non-contiguous optical coupler thereby avoiding the drawbacks associated with a contiguous optical coupler layer. Each of the processes produces a CT detector wherein a light transmission cavity is formed to collimate light emissions from a scintillator to a photodiode. The cavity may be filled with optical coupling epoxy or left empty and the scintillator bonded to the photodiode array. It is preferred that the cavities be filled with epoxy as this results in better optical transmission and a stronger connection being formed between the scintillator and photodiode.

Referring now to FIG. **11**, package/baggage inspection system **160** includes a rotatable gantry **162** having an opening **164** therein through which packages or pieces of baggage may pass. The rotatable gantry **162** houses a high frequency electromagnetic energy source **166** as well as a detector assembly **168**. A conveyor system **170** is also provided and includes a conveyor belt **172** supported by structure **174** to automatically and continuously pass packages or baggage pieces **176** through opening **164** to be scanned. Objects **176** are fed through opening **164** by conveyor belt **172**, imaging data is then acquired, and the conveyor belt **172** removes the packages **176** from opening **164** in a controlled and continuous manner. As a result, postal inspectors, baggage handlers, and other security personnel may non-invasively inspect the contents of packages **176** for explosives, knives, guns, contraband, etc.

Therefore, in accordance with one embodiment of the present invention, a CT detector includes a plurality of scintillators arranged in an array to receive x-rays and output light in response to the received x-rays. A plurality of light detection elements are arranged in an array dimensionally similar to the scintillator array and are configured to detect light from the scintillators. A non-contiguous optical coupler is then used to secure the plurality of scintillators to the plurality of light detection elements.

According to another embodiment of the present invention, a CT system includes a rotatable gantry having a bore centrally disposed therein and a table movable fore and aft through the bore and configured to position a subject for CT data acquisition. A high frequency electromagnetic energy projection source is positioned within the rotatable gantry and configured to project high frequency electromagnetic energy toward the subject. The CT system further includes a detector array disposed within the rotatable gantry and configured to detect high frequency electromagnetic energy projected by the projection source and impinged by the subject. The detector array includes a plurality of scintillators arranged in a scintillator array as well as a plurality of photodiodes arranged in a photodiode array. A light collimator having a plurality of light transmission cavities is disposed between the scintillator array and the photodiode array.

In accordance with a further embodiment of the present invention, a method of CT detector manufacturing includes the steps of forming a scintillator array having a plurality of scintillators and forming a photodiode array having a plurality of photodiodes. An open-celled collimator is then deposited between the arrays. The resulting assembly is then secured to one another.

The present invention has been described in terms of the preferred embodiment, and it is recognized that equivalents, alternatives, and modifications, aside from those expressly stated, are possible and within the scope of the appending claims.

What is claimed is:

1. A method of manufacturing a CT detector comprising the steps of:

forming a scintillator array having a plurality of scintillators;

forming a photodiode array having a plurality of photodiodes;

depositing an open-celled collimator between the scintillator array and the photodiode array, the open-celled collimator having collimator plates extending in a direction generally parallel to x-ray transmission and having a width perpendicular to the direction generally parallel to x-ray transmission that substantially matches a width separating adjacent scintillators; and

securing the scintillator, the open-celled collimator, and the photodiode array to one another.

2. The method of claim **1** wherein the steps of forming the scintillator array and depositing an open-celled collimator include the steps of:

mounting a block of scintillator material onto a dissolvable substrate;

dicing through the block and the dissolvable substrate to form a plurality of voids;

casting reflector material into the plurality of voids; and

dissolving the dissolvable substrate to form the open-celled collimator.

3. The method of claim **2** further comprising the step of filling each cell of the open-celled collimator with an optical coupling substance.

4. The method of claim **2** wherein the step of securing the scintillator, the open-celled collimator, and the photodiode array includes the step of coupling a thin layer of adhesive to the open-celled collimator.

5. The method of claim **1** wherein the steps of forming the scintillator array and depositing the open-celled collimator include the steps of:

mounting a block of scintillator material onto a thermoplastic substrate;

dicing through the block of scintillator material and at least partially through the thermoplastic material to form a plurality of voids; and

casting reflector material into the plurality of voids.

6. The method of claim **1** wherein the step of depositing an open-celled collimator includes the step of depositing a thermoplastic material with an embedded mesh between the scintillator array and the photodiode array, the mesh defining a plurality of light transmission cavities.

7. The method of claim **6** further comprising the step of filling the plurality of light transmission cavities with an optical epoxy.

8. The method of claim **1** wherein the step of depositing an open-celled collimator further includes the steps of:

coating the plurality of photodiodes with a semiconductor material; and

etching the semiconductor material to form a plurality of light transmission cavities.

9. The method of claim **8** further comprising the step of filling the plurality of light transmission cavities with optical epoxy.

10. The method of claim **1** wherein the open-celled collimator includes a metallic grid having a plurality of light transmission cavities.

11. The method of claim **10** further comprising the step of etching the metallic grid from a substrate.

12. The method of claim **10** wherein the step of securing includes the step of bonding the metallic grid to the plurality of scintillators and the plurality of photodiodes.

13. The method of claim **10** further comprising the step of filling the plurality of light transmission cavities with optical epoxy.

14. A method of constructing a CT detector comprising the steps of:

forming a scintillator array;

forming a photodiode array; and

securing the scintillator array to the photodiode array with an open-celled collimator, the open-celled collimator arranged to reduce optical cross-talk between adjacent scintillator-photodiode pairs.

15. The method of claim **14** wherein the steps of forming the scintillator array and securing the open-celled collimator include the steps of:

mounting a block of scintillator material onto a thermoplastic substrate;

dicing through the block of scintillator material and at least partially through the thermoplastic material to form a plurality of voids; and

casting reflector material into the plurality of voids.

16. The method of claim **14** wherein the open-celled collimator is formed by:

coating the photodiode array with a semiconductor material;

etching the semiconductor material to form a plurality of light transmission cavities; and

filling the plurality of light transmission cavities with optical epoxy.

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